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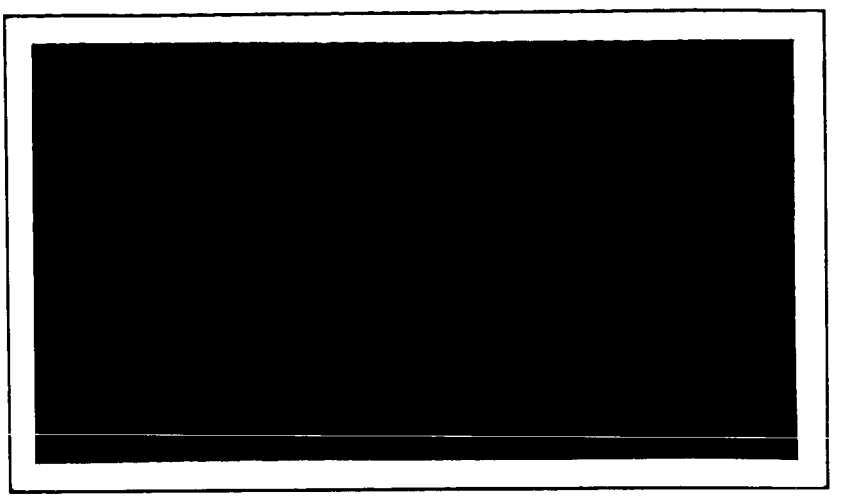

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BIOLOGICAL PASSIVE TELEMETRY

by

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Farmingdale, L. I., N. Y.

CONTENTS

ILLUSTRATIONS

Biological Passive Telemetry*

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27314 ABSTRACT:

Summary -- The main kinds of active and passive biological-telemetry schemes are described. A detailed examination of a scheme for passive biological telemetry is given. This method uses an implanted device that consists of an RF-tuned circuit. The resonant frequency of this circuit is controlled by the physiological parameter of interest, e.g., a temperature-sensitive capacitor (for temperature telemetry), a pressure-sensitive inductor (for pressure telemetry), or a voltage-sensitive capacitor (for biological-voltage telemetry). An external transmitter can be arranged to illuminate the implanted device with RF power, and in conjunction with an external receiver, the resonant frequency of the implanted device can be determined or the points in time at which variations occur. Factors affecting operation are examined, including RF attenuation in tissue, the types of fields that can be used, frequency considerations, coupling and directionality considerations, and techniques for the telemetry of biological voltage, temperature, pressure, and pH. The size of such a device and a discussion of the factors that might permit a reduction in this size also are given.

INTRODUCTION

The telemetry of physiological variables such as voltage, pressure, temperature, and pH, in recent years has been the subject of much study [1-13]. As medical and biological research progresses, it becomes increasingly important to develop tools for observing internal biological phenomena with a minimum of disturbance to the subject. In this paper, we consider an implantable passive telemetry device that would permit minimum disturbance of the subject and the possibility of observing physiological variables indefinitely. Minimum disturbance can be achieved by making the physical size of the implanted device extremely small. This would minimize tissue reaction and would reduce spurious effects in the physiological information that is obtained from the implanted device. A long-lifetime requirement dictates the use of passive telemetry techniques. Range of operation is a secondary consideration, provided that reliable operation is possible.

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** Research Division, Republic Aviation Corporation, Farmingdale, Long Island, New York.

Conversion to electrical power, using piezoelectric materials or a moving coil in a magnetic field, might be possible.

Active RF-Powered Transmitters. This type converts externally applied RF into a dc voltage that powers the transmitter. The operation may consist of an external transmitter at one frequency and an implanted device telemetering simultaneously at another frequency [15], or an external transmitter that charges a capacitor and, during the time when the transmitter is off, the charged capacitor powers the implanted oscillator for several cycles. In the latter case, the time between the end of the transmitted RF power and the beginning of transmitted power from the implanted device is made to be a function of a physiological parameter [12].

Passive Pulsed-RF Transmitters. This type also uses an RF-tuned circuit with the frequency of the implanted device being controlled by a physiological parameter. RF power is periodically supplied from an external transmitter. The frequency of this transmitted RF-pulse power lies somewhere in the vicinity of the resonant frequency of the implanted device. During the interpulse time, the frequency of the decaying RF currents in the implanted device are measured at an external receiver. This method has been applied to the measurement of intragastric pressure [10,11].

Passive RF-Powered Transmitters. This type uses an implanted device consisting of an RF-resonant circuit. One of the components (L or C) of the resonant circuit is made to be a function of the physiological parameters of interest, e.g., a pressure-sensitive inductor, a temperature-sensitive capacitor, or a voltage-sensitive capacitor. The resonant frequency is controlled by the physiological parameter. The implanted device is operated by an external transmitter/receiver in a way similar to that of a grid-dip meter; when the frequency of the transmitter is equal to the resonant frequency of the implanted device, then, depending on the Q of the implanted device, RF currents will flow in the device. A receiver will detect this current in such a way that it can be differentiated from the transmitted power. This paper is a detailed study of this means of passive telemetry (see Fig. 1).

PHYSIOLOGICAL PARAMETERS

The use of implanted or even externally attached telemetry devices should be dictated by necessity. When used, much useful information can be gathered

The different types of implantable telemetry devices that have been developed usually differ in their method of obtaining power for operation. The characteristics of these several types are summarized in the following paragraphs.

Active Battery-Powered Transmitters. A great deal of success has been achieved with small battery-powered FM transmitters [1-6, 9]. These devices usually transmit an FM signal from within the subject to an FM receiver external to the subject. The frequency deviation of the transmitter is usually made a function of the magnitude of the physiological parameter of interest. The source of power is mercury batteries because of their high energy-to-volume ratio. When short-time applications are required (such as the telemetry of pressures in the digestive tract from a swallowed "radio pill"), this type is most useful. In addition, for work with birds and small animals, ranges of many miles are possible. Its limited lifetime in implanted applications has stimulated work on external means of switching the transmitter on and off to conserve power and on the coupling of RF power into the telemetry transmitter to recharge the batteries. These methods, however, have a tendency to increase the volume of the implanted device.

Active Biochemically-Powered Transmitters. This type is similar to the one described above but it differs in that it obtains its power from the chemical or biochemical energy that is available in the subject. Recent experiments with mice using electrodes in the stomach and peritoneal cavity indicate that the reduction potential of the stomach acid can be used for the generation of microwatts of power [14]. This method requires, however, that some epithelial barrier be breached to make such power available. If electrode-polarization problems are solved, high power levels are available.

Active Mechanically-Powered Transmitters. This type converts mechanical energy in the subject to electrical energy to power a transmitter. Some work has been described [7] that utilizes the motion of muscles to bend or deform piezoelectric materials for such conversion. The expansion and contraction of the aorta artery can be harnessed by having a ring contain piezoelectric chips [8] situated around the aorta. This method makes available intermittent power so that it might be required that such energy be stored in batteries or else the physiological information will have to be transmitted intermittently. Mechanical energy also can be extracted from the gross motion of a subject using a self-winding watch mechanism. This energy could be stored in a coiled spring and made available through an escapement.

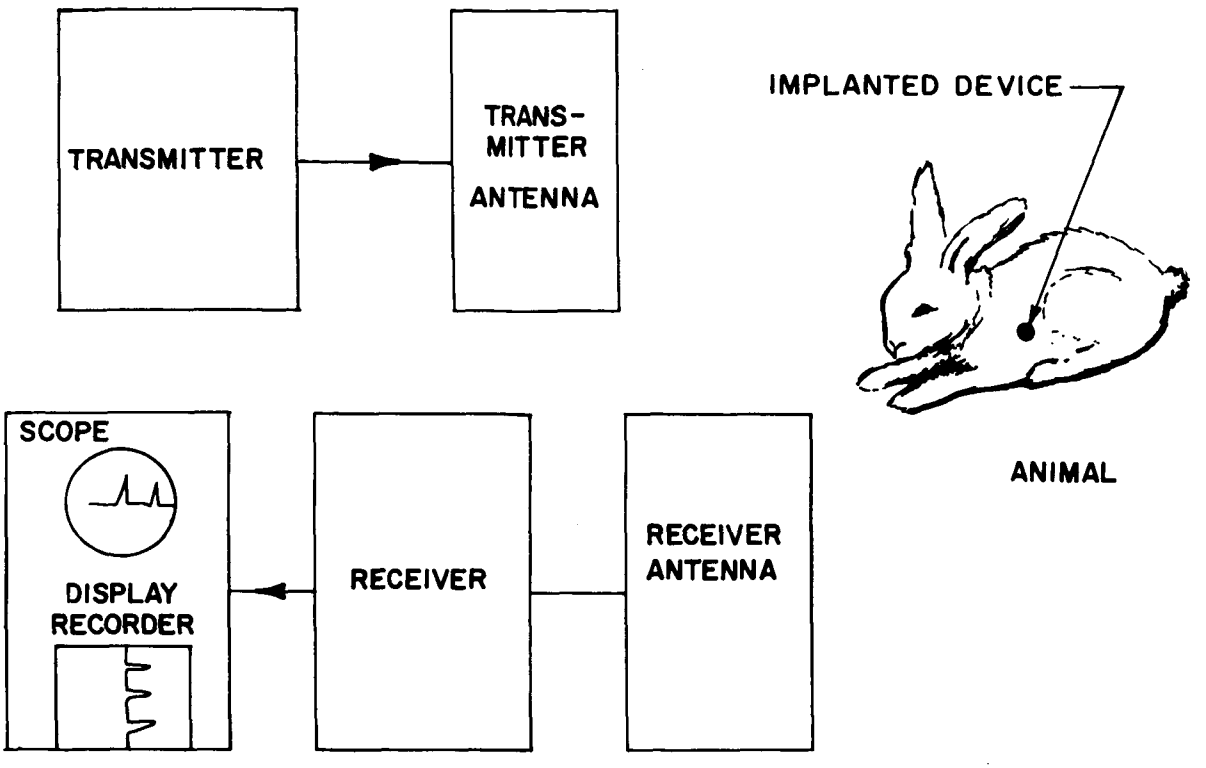

Figure 1. Block Diagram of Passive-Telemetry System

quickly and with a minimum of difficulty by wiring the output of biological sensors directly to display or recording devices [16]. In the discussion that follows, only passive, RF-powered-transmitter telemetry will be considered. In this case, the physiological parameter must affect the resonant frequency of the RF-tuned circuit in the implanted device.

NERVE AND MUSCLE POTENTIALS

The telemetry of nerve or muscle voltages requires that some estimate be made of the equivalent-circuit parameters of the tissue. Electrodes are necessary to utilize these voltages for changing the resonant frequency of the passively tuned circuit. Initially, polarization effects at the electrodes are neglected.

A nerve cell may be considered as a transmission line. The nerve membrane separates the interior of the nerve from its environment. Biochemical processes maintain the interior of the cell at a negative potential with respect to the exterior. This voltage (about -70 mv) is the same for most animal species, including man. Nerve signals consist of abrupt reversals of this potential (to about +30 mv) that

move down the nerve at speeds of 1-100 m/sec. After the passage of this signal, metabolic processes cause a rapid recovery to the resting potential (-70 mv).

The abrupt reversal of the potential is called an action potential spike and is an all or none response in that such waveforms are standard in magnitude and duration and will propagate down the cell, but lower voltages will not. The duration is in the range of 1-6 msec, and from 5 to 500 spikes per second might pass down a cell fiber. The action potential, with its peak magnitude of +30 mv, may be represented in an equivalent circuit by a transient voltage source of about 100 mv. Estimates of the current flowing through nerve cells are of the order of 10^{-8} to 10^{-9} amp [17-19].

The diameter of nerve cells varies from 1μ in some warm blooded animals to 1mm in some squid. The variation in most warm blooded animals is between 1-100 μ , but 10 μ will be assumed to be the average diameter for the purpose of calculation. The length of nerve cells varies widely; they can be 4-5 ft in man.

A single cell may be replaced by the equivalent circuit of Fig. 2a during conduction, and the internal resistance R of the cell can be found by

$$R = V/_i = 0.05/5 \times 10^{-9} = 10^7 \text{ ohm} \qquad (1)$$

where V is the output voltage of the cell when loaded for maximum power transfer and i is the current in the cell.

On the basis of estimates of cell size [17, 18], it is assumed that an average cell may be represented as a cube 10 μ on edge. The electrode that contacts the tissue is assumed to have an area of 1 mm^2. Thus, the electrode will contact the number of cells that can fit this area or 10^4 cells (see Fig. 2b). This estimate also may be used if one assumes that the nerve tissue consists of 1-μ -diameter, parallel fibres with lengths averaging 100 μ contacting the electrode.

A common type of coding for signals that are transmitted along nerve cells is one in which the number of action potential spikes per second is a function of the magnitude of the signal that the organism is transmitting from one point to another. The signals passing down separate but contiguous nerve cells are usually not synchronous, that is, the action potential spikes do not occur at the same point in the tissue or in time. This condition, however, does occur in heart tissue or, during physical activity, in muscle tissue. Although the structure of muscle tissue is not the same as that of nerve tissue, many of the previous considerations apply to both.

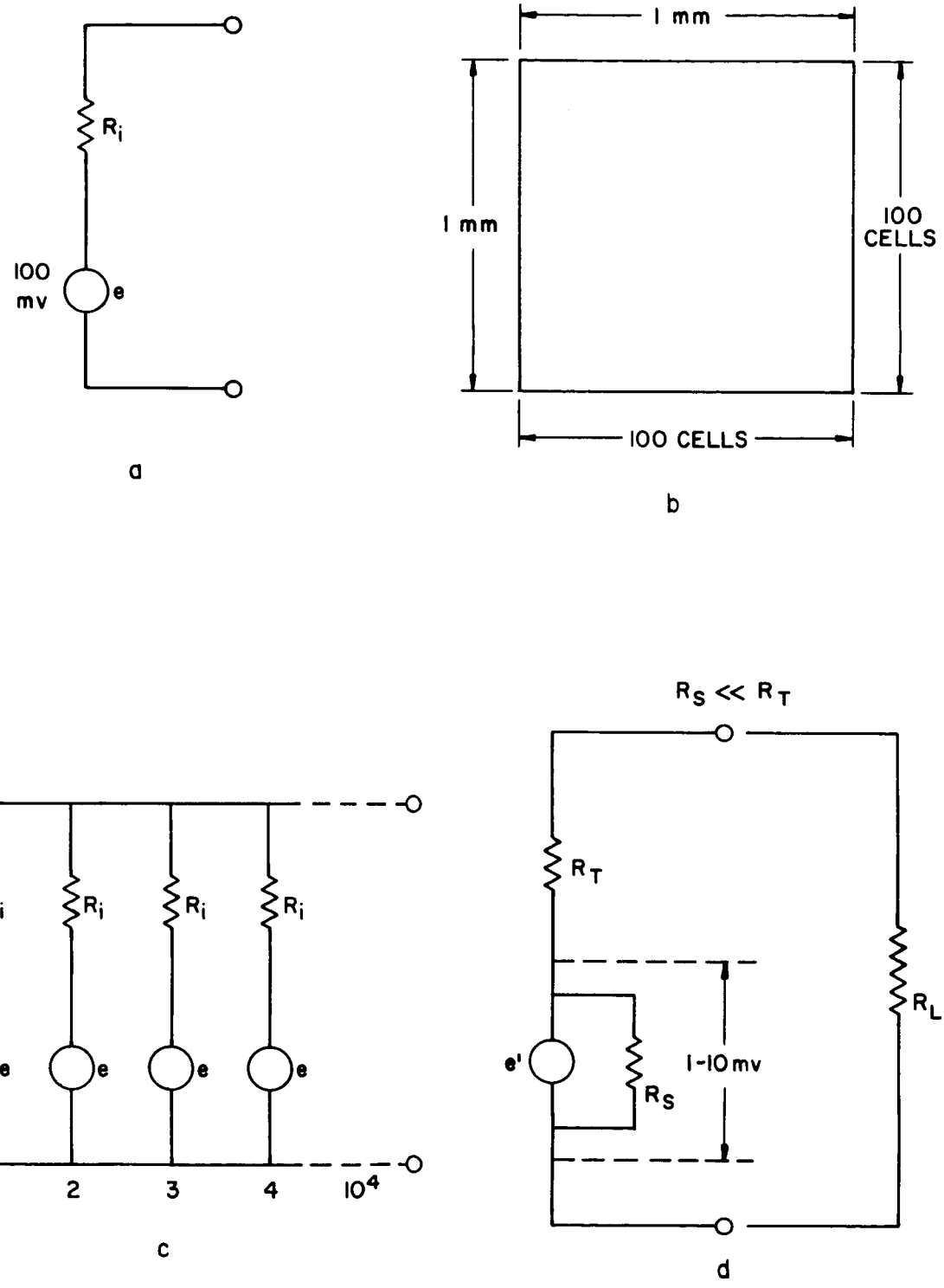

Figure 2. Tissue-Equivalent Circuits

One may draw an equivalent circuit for the case of 10^4 cells that are in contact with an electrode having an area of 1 mm^2. This consists of 10^4 equivalent circuits, like those shown in Fig. 2c, in parallel, each equivalent circuit corresponding to each cell. One may use Thevenin's theorum to derive a single equivalent circuit (see Fig. 2d) corresponding to this case in which:

$$R_t = R_i/N = 10^7/10^4 = 1000 \text{ ohm} \qquad (2)$$

where R_t is the equivalent internal resistance for the total number of cells N. These equivalent circuits can be applied particularly to the case of heart and muscle tissue.

Although all the cell-equivalent circuits are assumed to be in parallel with the electrode, this is not actually the case. The equivalent circuit for the tissue that the electrode will contact must be modified for the presence of leakage paths in the tissue that reduce the voltage measured at the electrode. This modification also must take into account the fact that the electrode will be measuring the voltages on the surfaces of the cells.

The voltages are measured with reference to the tissue or fluid that is not being traversed by action potential spikes. The tissue voltage at the electrode was estimated to be in the range of 1-10 mv. Actual measurements of this voltage and of the impedance (1,000 ohm) in the heart tissue of turtles and dogs tend to confirm this estimate.* Thus, for electrode areas of 1 mm^2, 10^4 cells are cooperating to deliver about 10^{-7} watts at the peak of an action potential spike. If the action potential spikes are used to vary the resonant frequency of a tuned circuit by means of a voltage-variable capacitor, then no more than the above power can be used in such an arrangement.

In the telemetry of data on action potential spikes, one usually need not be concerned with the accurate reproduction of the spike waveform as a function of time because it is the number of spikes per second and the time of occurrence that is of interest. Nevertheless, several methods of preserving the fidelity of the waveform during telemetering are described later in this paper.

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* Tissue voltage and impedance measurements as a function of electrode area were made on the heart tissue of living dogs in August 1962 at Downstate Medical Center by Dr. B. Hoffman and Dr. J. F. Stuckey. Similar measurements on turtle hearts were made at Republic in October 1963.

PRESSURE

Intragastric pressure has been used to vary the magnitude of an inductance of a tuned circuit [11]. Pressure on a piezoelectric chip (to vary the capacitance of a voltage-variable capacitor) also can be used to vary the resonant frequency of a tuned circuit [4]. This method can be used for the measurement of blood pressure.

BODY TEMPERATURE

For the passive telemetry of temperature, a temperature-sensitive capacitor in the RF-tuned circuit can be used. Such an implanted device should be small in size and have a low thermal mass so that there is less disturbance of the subject and so that temperature artifacts be minimized.

HYDROGEN-ION CONCENTRATION

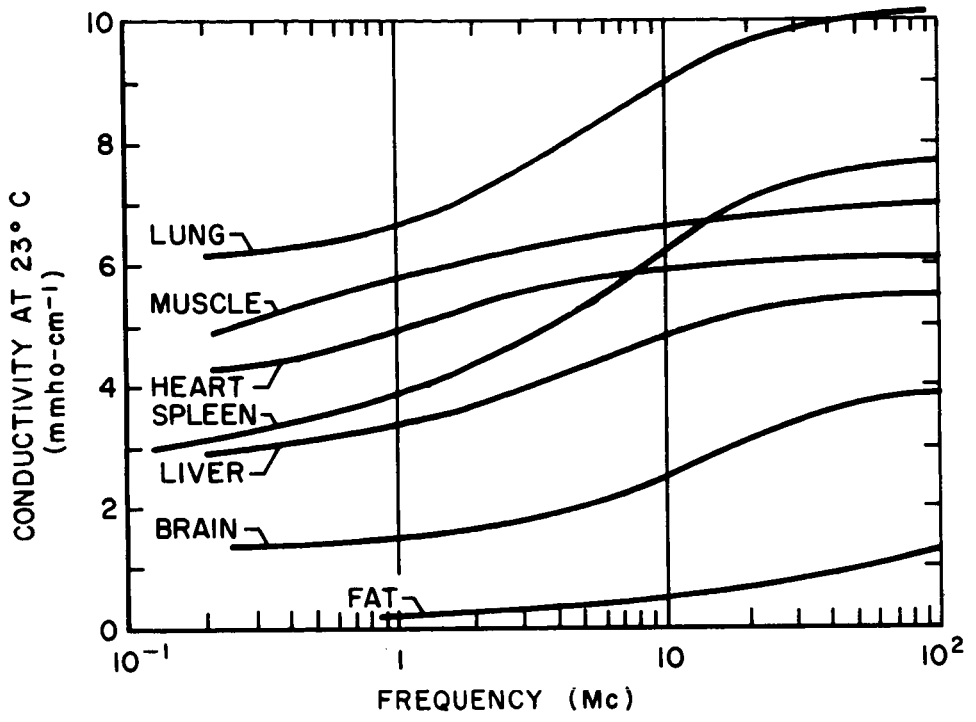
The measurement of pH usually is accomplished with a set of leads the output voltage of which is a function of pH. This output can be used to vary the capacitance of a voltage-variable capacitor in an RF-tuned circuit. This method has been used in active devices to vary the frequency of a transmitter [4].

PASSIVE-TELEMETRY CONSIDERATIONS

PROPAGATION

Many studies of the RF properties of biological material have been made [15, 20, 21]. A plot of the conductivity and a plot of the dielectric constants of several tissues as a function of frequency are shown in Fig. 3 [20, 21]. It will be shown that most considerations of the propagation of RF through tissue for passive telemetry will be in the 5-100 Mc range.

Table 1 lists the values of conductivity and dielectric constant for sea water and for "average" tissue. The tissue values were determined from Fig. 3 over the 5-100 Mc range. Although such procedures are unreliable for detailed applications, they provide an indication of the variation of losses with frequency. Sea water is included because initial propagation tests of telemetry devices were carried out in sea water, using it as a substitute for tissue.



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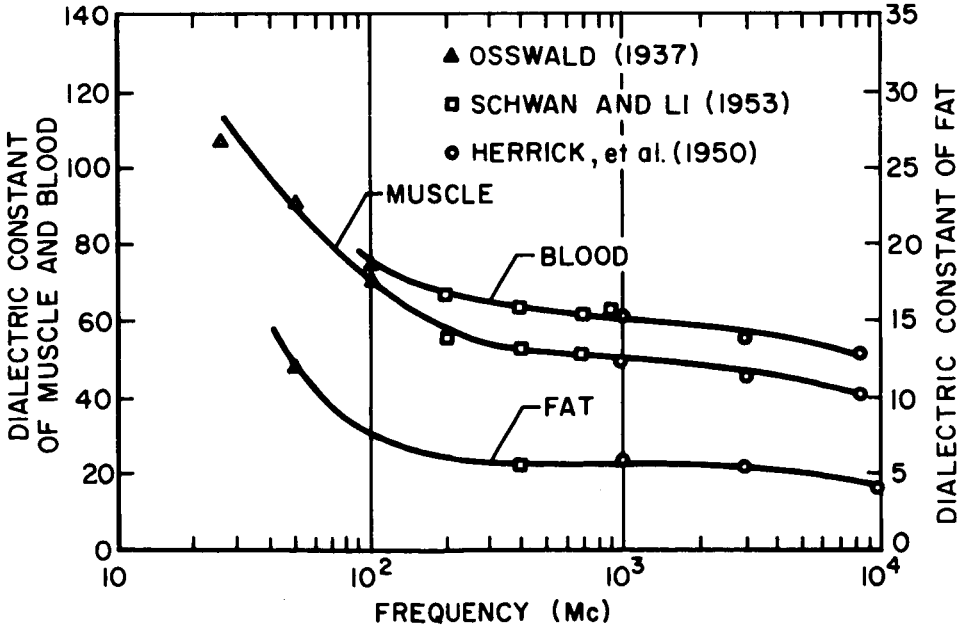


b.

Figure 3. a) Conductivity of Several Tissues
b) Dielectric Constant of Several Tissues

TABLE 1

	Conductivity (mho/centimeter)	Dielectric Constant
Sea water [23]	0.05	80
"Average" tissue	0.005	80

Following Smythe [22], the propagation constants for electromagnetic waves in a homogeneous conducting medium are

$$n^2 = \frac{\mu K}{2}\left[\left(1+\frac{16\pi^2}{\omega^2\tau^2K^2}\right)^{1/2}+1\right] \tag{3}$$

and

$$k^2 = \frac{\mu K}{2}\left[\left(1+\frac{1+16\pi^2}{\omega^2\tau^2K^2}\right)^{1/2}-1\right] \tag{4}$$

where

n = index of refraction
k = coefficient of extinction
μ = permeability
K = dielectric constant
ω = $2\pi f$
f = frequency
τ = resistivity $= 1/\sigma =$ 1/conductivity

For the case in which $K \cong 80$ and τ is as shown in Table 1 for sea water and "average" tissue, the value of the square-root terms in Eqs. 3 and 4 for 5, 50, and 100 Mc is shown in Table 2.

TABLE 2

	5 Mc	50 Mc	100 Mc
Sea water	>6,800	>68	17.1
"Average" tissue	> 430	>4.3	1.08

When the square-root terms are large with respect to one

$$n \cong k \cong \left(\frac{\mu\lambda_o}{c\tau}\right)^{1/2} \tag{5}$$

where:

λ_o = free-space wavelength

c = velocity of light

This is true for the cases shown in Table 2 except for "average" tissue at 100 Mc and greater. In this case, the value of k is below that given by Eq. 4.

The electric and magnetic-field vectors will contain the term $e^{-\omega kd}$. The distance d that an electromagnetic wave travels until the field vector falls to $e^{-2\pi}$ of its initial value is

$$d = \left(\frac{c \tau \lambda_0}{\mu}\right)^{1/2} \tag{6}$$

A plot of the depth of penetration for an 8.7-db loss in the electromagnetic wave is shown in Fig. 4 for sea water and "average" tissue.

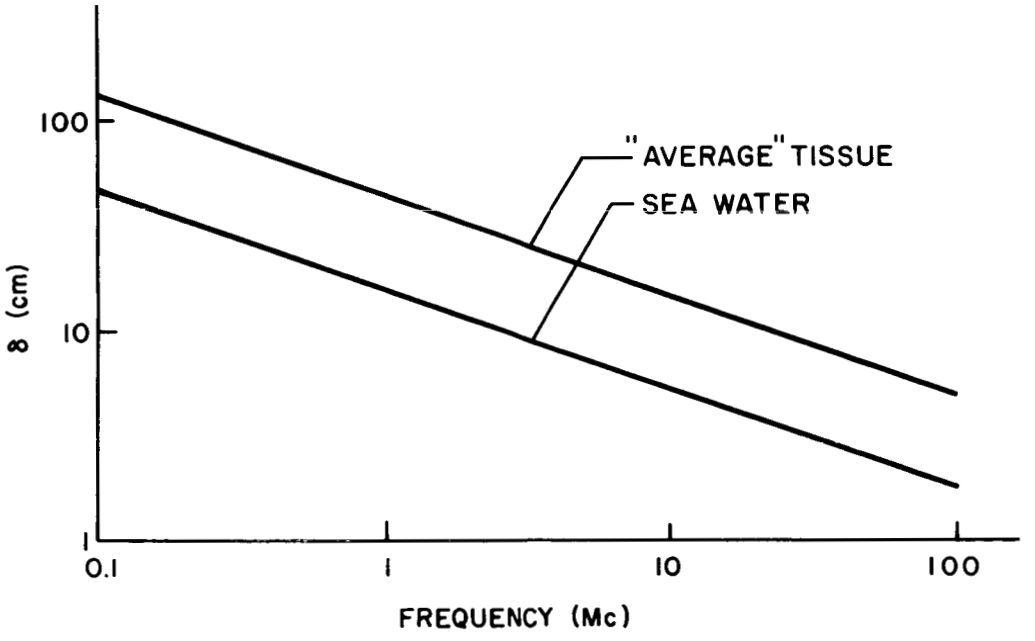


Figure 4. Depth of Penetration for 8.7-db Loss

The ratio of the average electric to magnetic energy is

$$\frac{W_e}{W_m} \cong \frac{Kc\tau}{2\lambda_0} \tag{7}$$

For the sea water and "average" tissue, this ratio is as shown in Table 3.

TABLE 3

Frequency (Mc)	Ratio of Average Electric to Magnetic Energy	
	Sea Water	"Average" Tissue
100	0.228	0.912
10	0.0228	0.0912
1	0.00228	0.00912

CIRCUIT ANALYSIS

For the case in which a transmitter, implanted device, and receiver are coupled by means of the induction fields of their coupling coils, a circuit analysis can be made (see Fig. 5). For this analysis, we will use subscripts 1 for the transmitter loop, subscripts 2 for the implanted device, and subscripts 3 for the receiver.

Thus, one may write

$$\begin{aligned} E &= A_1 I_1 + B_1 I_2 + B_3 I_3 \\ 0 &= B_1 I_1 + A_2 I_2 + B_2 I_3 \\ 0 &= B_3 I_1 + B_2 I_2 + A_3 I_3 \end{aligned} \tag{8}$$

where

$$A_i = R_i + j\omega L_i - \frac{j}{\omega C_i} \quad (i = 1, 2, 3) \tag{9}$$

$$B_i = j\omega M_i \quad (i = 1, 2, 3) \tag{10}$$

where M is the mutual inductance between each component.

The transmitter and receiver inductors can be positioned for minimum coupling so as not to saturate the receiver. The coupling of energy directly from transmitter to receiver is not of interest. A CW-transmitter signal is assumed, and this signal may be ignored when solving for V_o. In addition, the coupling from transmitter to implanted device and from receiver to implanted device are made to be equal, and the parameters of the transmitter- and receiver-output circuits are assumed to be equal, so that $A_1 = A_3$, $B_3 = 0$, $B_1 = B_2$, $V_o = I_3 \left(\frac{j}{\omega C_3}\right)$ or $V_o = I_3 (R_3 + j\omega L_3) \cong I_3\, j\omega L_3$, where

$$I_3 = \frac{E\, B_1^2}{\Delta} \tag{11}$$

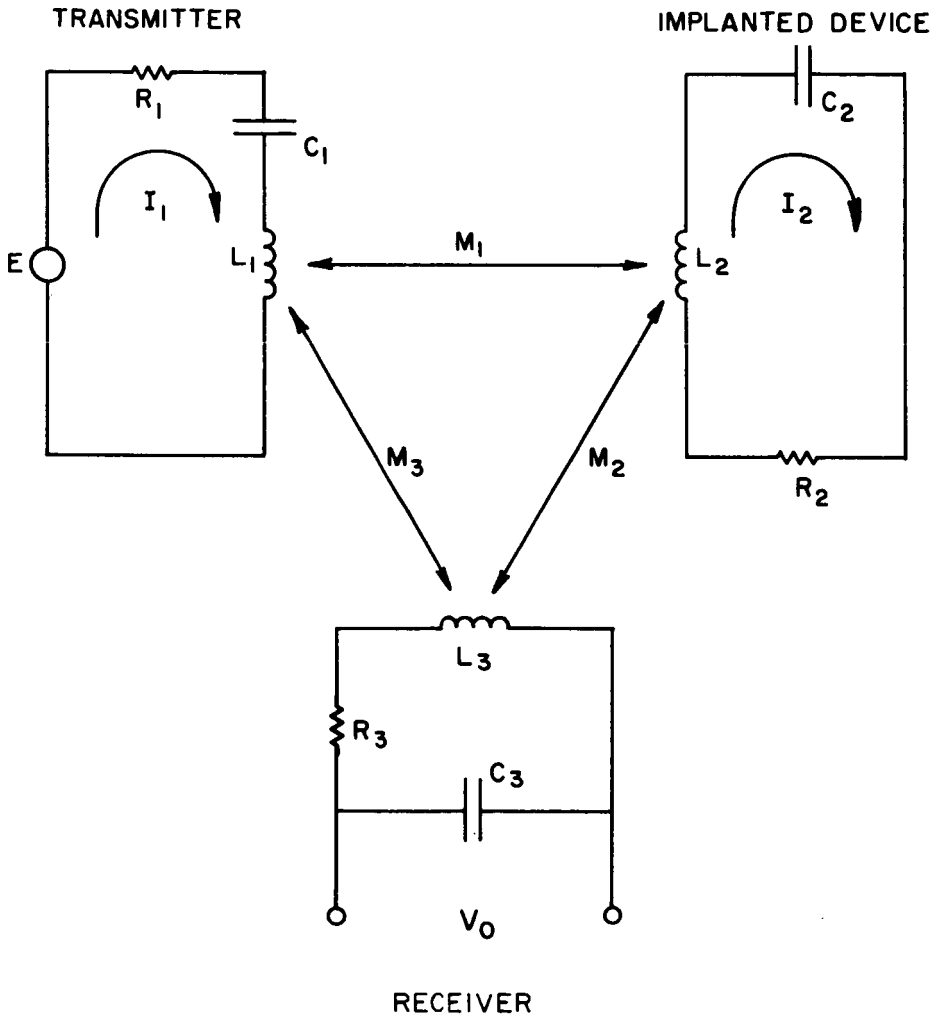


Figure 5. Schematic of Passive-Telemetry System

and

$$\Delta = A_1^2 A_2 - 2A_1 B_1^2 \tag{12}$$

For $B_1 << A_1$ or A_2, as it might be in an actual case

$$I_3 = \frac{E}{A_1}\left(\frac{B_1^2}{A_1 A_2}\right) \tag{13}$$

and

$$V_o = \frac{E}{A_1}\left(\frac{B_1^2}{A_1 A_2}\right) \quad j\omega L_3 \tag{14}$$

At resonance $A_i = R_i$, and

$$Q_i = \frac{\omega L_i}{R_i} \quad (i = 1, 2, 3) \tag{15}$$

or

$$R_i = \frac{\omega L_i}{Q_i} \quad (i = 1, 2, 3) \tag{16}$$

In the region of resonance, therefore

$$\left|V_o\right| = E\left(\frac{M_1^2}{L_1^2}\right) \frac{L_3 Q_1^2 Q_2}{L_2} \tag{17}$$

For the special case of two coaxial coils the separation of which is large compared to the smaller of the coils [24]

$$M_1 = p \frac{a_1^2 a_2 n_1 n_2}{r^3} \tag{18}$$

and

$$V_o = \frac{qE}{r^6} \tag{19}$$

where

p, q = proportionality constants

a_i = radius of coil

n_i = number of turns of coil

r = separation in axial direction (assuming that receiver and transmitter are at the same location)

Note that the voltage at the receiver will vary inversely as the sixth power of the separation between the coils with the above assumptions and approximations.

DIRECTIONAL CONSIDERATIONS

The coupler inductors of transmitter, receiver, and implanted device obviously cannot be coaxial if the receiver and transmitter coils are positioned for minimum coupling. The receiver signal V_o will be sensitive to the position of the implanted inductor L_2 with respect to L_1 and L_3, assuming that the receiver and transmitter inductors are positioned for minimum coupling.

Directionality considerations can be minimized by assuming that the transmitter causes an RF magnetic field in the implanted device (see Fig. 6). Thus, when the mutually perpendicular inductors L_x, L_y, and L_z create the x, y, and z magnetic fields

$$H_T = H_o (\cos \alpha \hat{i} + \cos \beta \hat{j} + \cos \gamma \hat{k}) \qquad (20)$$

where

H_T = magnetic-field vector of implanted device

H_o = magnetic-field vector of transmitter at implanted device

$\hat{i}, \hat{j}, \hat{k}$ = unit vectors in x, y, and z directions

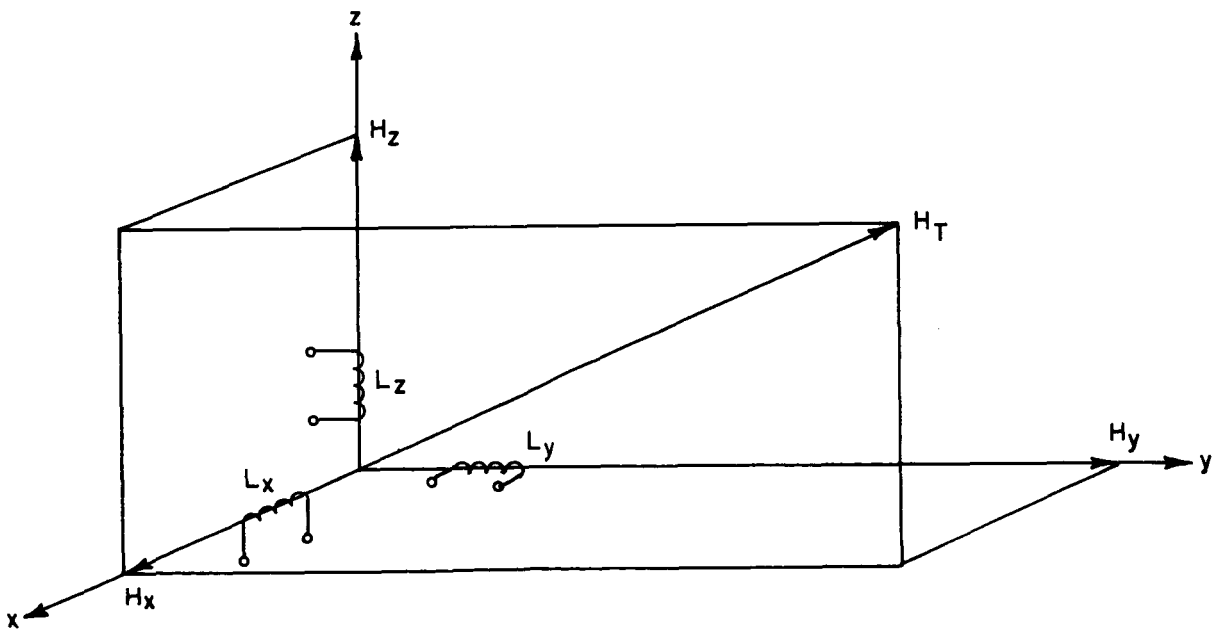


Figure 6. Directional Considerations

These inductors may be connected in series or parallel combination and resonated with a capacitor.

Further, assuming that magnetic interaction between inductors is slight, then

$$\begin{aligned} |V_x| &= qH_x = qH_o \cos\alpha \\ |V_y| &= qH_y = qH_o \cos\beta \\ |V_z| &= qH_z = qH_o \cos\gamma \end{aligned} \tag{21}$$

where V_x, V_y, and V_z are the voltages available at the terminals of L_x, L_y, and L_z and q is a proportionality constant. Also

$$V_x^2 + V_y^2 + V_z^2 = q^2 H_o^2 \left[\cos^2\alpha + \cos^2\beta + \cos^2\gamma\right] = q^2 H_o^2 \tag{22}$$

Thus, the sum of the voltages squared is independent of the orientation between the magnetic-field vector H_T and the three mutually perpendicular coils. To satisfy Eq. 22, however, would require that three mutually perpendicular transmitting coils and three mutually perpendicular receiver coils would have to be placed around the implanted device (and, hence, the subject) with the output of the receiver coils being fed to squaring circuits and then added [25]. Obviously, this method would be cumbersome.

In the actual system (see below), the transmitter coil is oriented so that direct coupling from transmitter to receiver is minimized, and the implanted device contains the two mutually perpendicular coils. These coils are not free from interaction, but they do not appear to cause as great a directional sensitivity as would a single coil.

In general, coupling between receiver, transmitter, and implanted device is caused by induction fields. The free-space wavelength in the 5-50 Mc region varies from 60 to 6 m. If the implanted device contained core materials such as the newer ferrites (i. e., Indiana General Q-2), then a dielectric constant of 10 and a permeability of 40 would give wavelengths in the ferrite materials of 1/20 of the above wavelength.

Wheeler [26] has shown that the ratio of the radiation resistance R_m to the inductive reactance ωL is the so-called radiation power factor p_e as given by

$$p_e = \frac{R_m}{\omega L} \cong \frac{1}{6\pi}\left(\frac{V}{\ell^3}\right) \tag{23}$$

where V is the volume of the antenna, and ℓ is the radian length of the antenna ($\ell = \frac{\lambda}{2\pi}$). Equation 23 applies to inductive or capacitive antenna structures that are small compared to the wavelength.

The magnitude of p_e is shown in Table 4 for an antenna consisting of a cube 0.15 in. on edge. At 50 Mc, the distance from such an antenna at which the magnetic vector in the electromagnetic field is equal to the magnetic-induction vector is greater than several meters (N = 10 ft in air). At 10 and at 5 Mc, this critical distance is 100 and 300 times greater than this value, respectively (1,000 and 10,000 ft). Therefore, although the induction fields fall off as $1/r^3$ and the electromagnetic fields fall off as $1/r^2$, initial tests in the 5-10 Mc region indicate that ranges of several feet are possible. The magnetic-induction field, therefore, is the major coupling mechanism. The use of a magnetic-induction field for the transfer of energy into animals has been described [27, 28].

TABLE 4

	5 Mc	10 Mc	50Mc
Free-space wavelength (m)	60	30	6
Wavelength in ferrite (m) (K = 10, μ = 40)	3	1.5	0.3
Radian length (m)	0.5	0.25	0.05
Radiation power factor	10^{-8}	10^{-7}	10^{-5}
Critical distance (ft)	>300	$> 10^2$	> 10

SYSTEM OPERATION

The passive telemetering of biological voltages can be achieved with an implanted device such as that shown in Fig. 7. The biological voltage is impressed on electrodes z and z′, where z is the tissue electrode with an area of 1 mm^2 and z′ is the ground or reference electrode with an area (at least three z) such

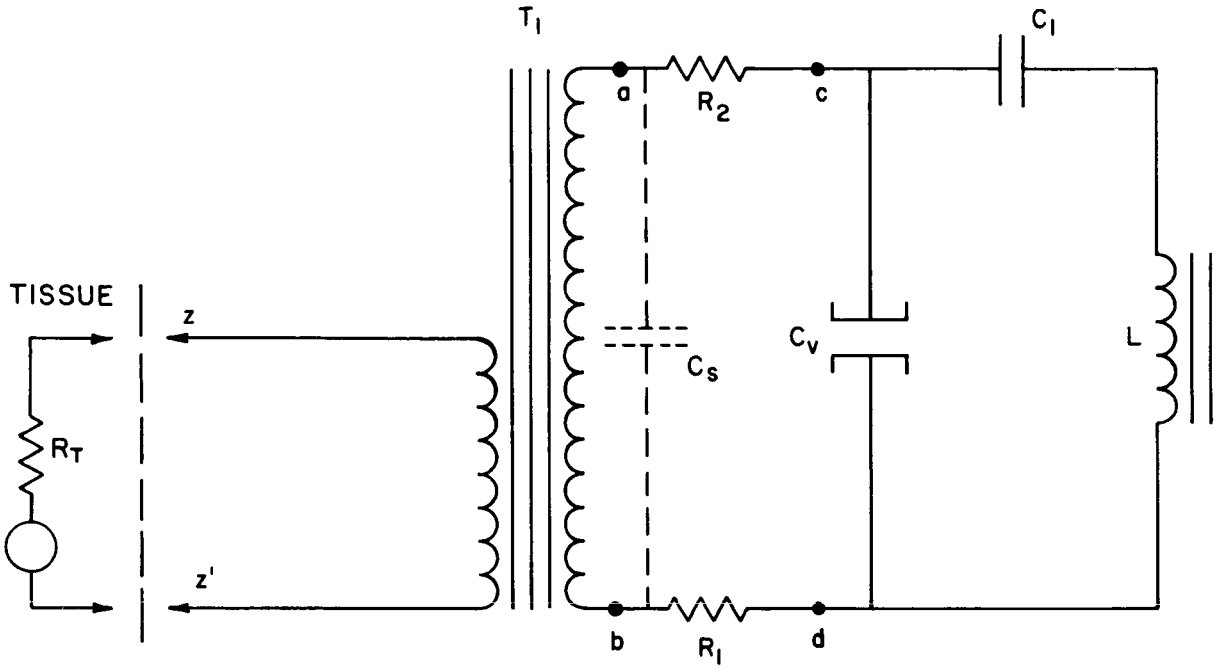


Figure 7. Schematic of Implanted Device

that the total tissue resistance R_T is about 1,330 ohm. If z and z' each had an area of 1 mm^2, then R_T would be 2,000 ohms.

The biological voltage is stepped up 100 times by transformer T_1. This stepped-up voltage is applied to the voltage-variable microjunction diode C_v. The resistance of this diode in its reverse direction is 10^9 ohm or more. The magnitude of the reactance of C_v is greater than 10^8 ohm when $C_v < 10\ \mu\mu f$. This means then the impedance to the right of points c and d is greater than 10^8 ohm at frequencies of 1,000 cps or less and that this impedance is 10^4 ohm or greater when reflected in the primary. This amount would not appreciably load the tissue. Significant loading occurs, however, because of the magnetizing reactance (see Appendix).

The resistors R_1 and R_2 are greater than 5,000 ohm. They do not affect the biological waveforms but are required to isolate the RF components C_v, C_1, and L from the secondary shunt capacitance of transformer T_1. The magnitude of the C_v, C_1 and L reactances at RF frequencies are less than the 5,000-ohm isolation between the transformer secondary-shunt capacitance C_s and the RF components effectively minimizes the loading of the RF components.

The operation of the telemetry system can be understood from an examination of the waveforms involved (see Fig. 8). The assembled characteristics of the biological voltage V_b are as follows:

Voltage amplitude	1-10 mv
Duration	1-20 msec
Interpulse period	.01-1 sec
Waveform rise time	0.3-1.5 msec

The biological voltage V_b will vary with time in some fashion like that of Fig. 8a. The output voltage of the transformer is equal to 100 V_b.

The capacitance C_v of the back-biased microjunction diode is a decreasing function of back-bias voltage (see Fig. 8b). This may be expressed as

$$C_v = p\,(V_w + V_1)^{-1/2} \tag{24}$$

where V_w is 0.65 v at 37°C, V_1 is the back-bias voltage and p is a proportionality constant. The variation of the capacitance as a function of time is shown in Fig. 8b for the voltage-variable capacitance C_v.

The resonant frequency of the RF tuned circuit is given by

$$\omega^2 L C_T = 1 \tag{25}$$

where L is the RF inductance and C_T is the total RF capacitance, i.e., the capacitance of C_v and C_1 in series. The resonant frequency of the tuned circuit will vary with time in response to the voltage across it in some fashion like that shown in Fig. 8c.

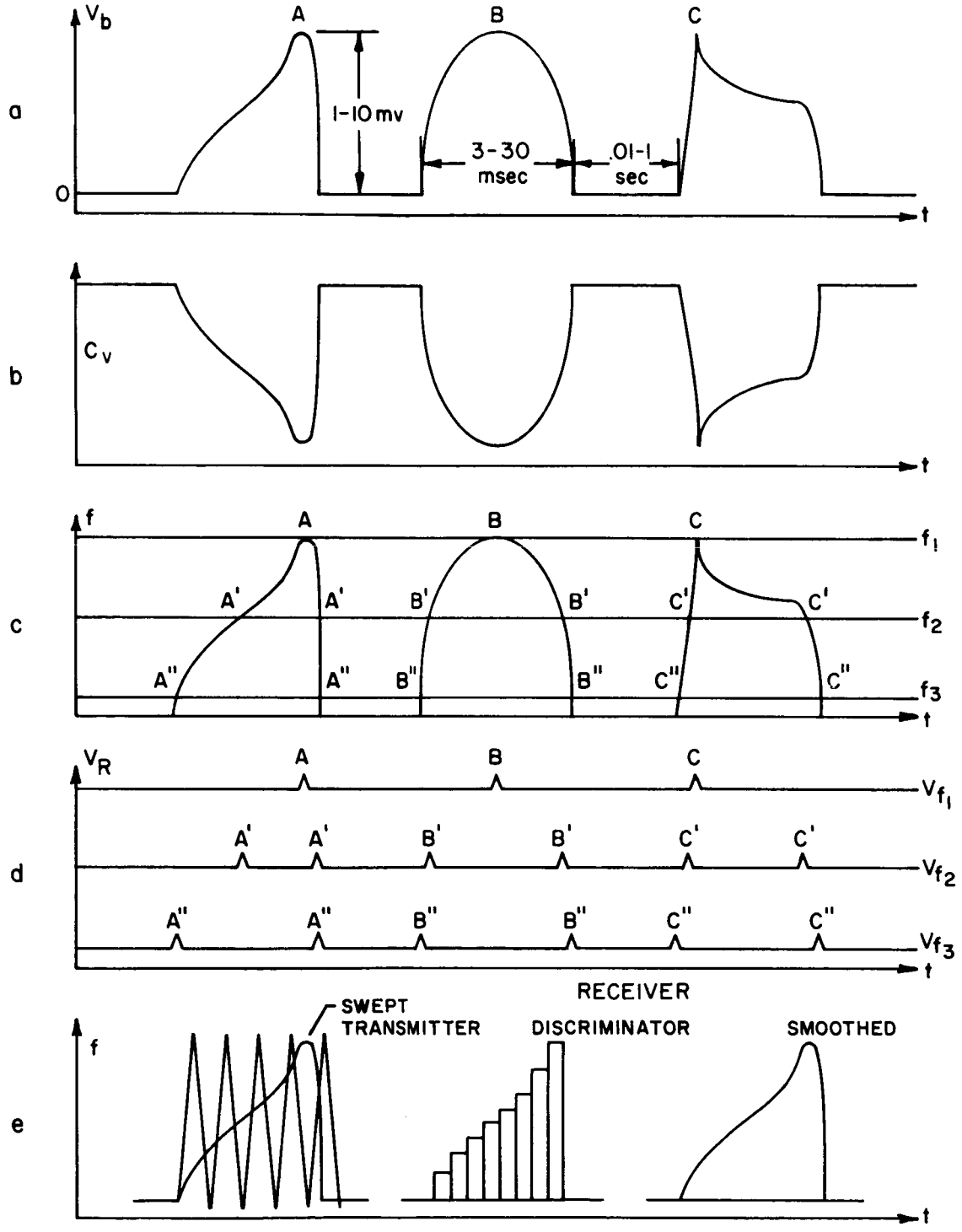


Figure 8. Telemetry Waveforms

There are several arrangements by which passive telemetry can be accomplished. Consider first a CW transmitter and a fixed, tuned receiver operating at frequency f (see Fig. 8c) situated near the implanted device so that voltage V_R exists at the receiver output terminals. Whenever the RF resonant frequency of the implanted device is swept through f_1 (points A, B, and C of Fig. 8c), there is a corresponding variation in the signal level of the receiver (points A, B, and C of Fig. 8d). If the operating frequency is set at f_2, then, when the RF resonant frequency of the implanted device sweeps through this value (points A′, B′, and C′ in Fig. 8c), there will be a variation in receiver voltage at points A′, B′, and C′ (Fig. 8d), and so on. Note that double indications occur at the receiver; these indicate the rise and fall of the biological voltages. Operation at the peak of the biological voltage gives just one indication.

For applications that require only the indication of presence or absence of biological pulses and for timing studies, this arrangement, using a CW transmitter and a fixed, tuned receiver, is adequate. At a single CW frequency, this arrangement can be varied so that reliable pulse-rise or pulse-fall indications can be monitored and then recorded with standard instrumentation. Because it is usually the number of pulses per second that is proportional to a stimulus and because amplitudes are generally coded by pulse number in nerve pathways, this arrangement might be adequate for a wide number of applications.

On the basis of the previous discussion, it might be inferred that only the telemetry of monopolar voltages are possible. Bipolar voltages also can be telemetered in one of the following three ways:

1) The curve of capacitance vs. back bias for C_V extends into the positive-voltage region for an amount somewhat less than V_W. This means that, for the case under consideration, if the bipolar voltage remains less than 6.5 mv, bipolar operation is possible as long as the resistance remains greater than 10^7 ohm.

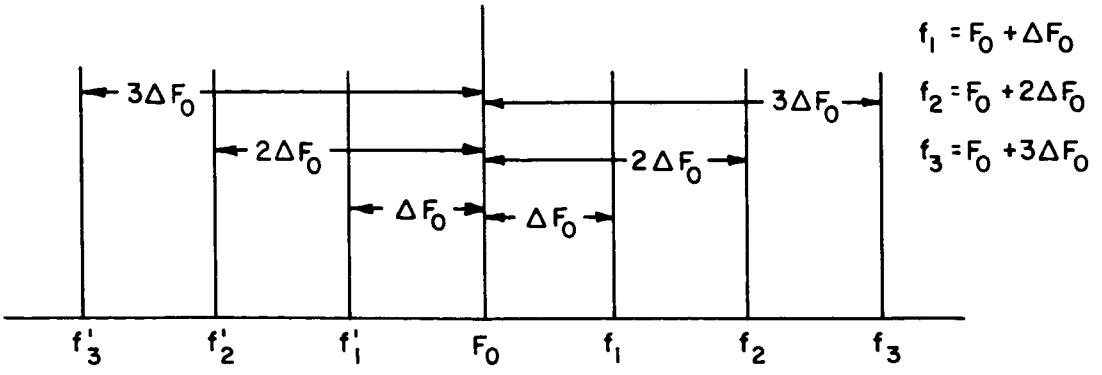
2) The electrodes z and z′, which are, respectively, "hot" and ground leads, govern, to some extent, the polarity of the biological voltages, i.e., the lead z usually will be attached to the tissue generating the biological voltages of interest, which are positive with respect the the common lead z′. (The shape of the implanted device can affect the polarity to some extent.)

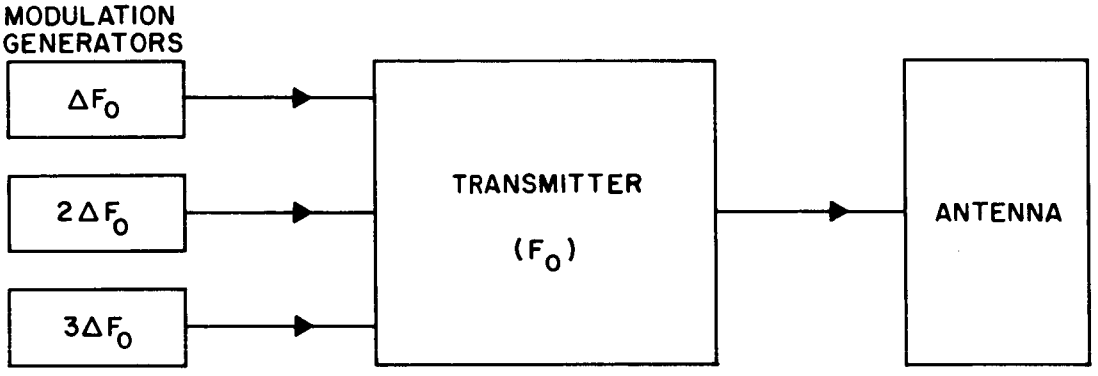
3) The secondary of transformer T_1 in Fig. 7 can be used to drive two separate RF-tuned circuits, each having oppositely connected reverse-bias diodes. The effect of this arrangement is as if there were two different, implanted devices, each telemetering one polarity of biological voltage.

When detailed information about the biological waveform is needed, a somewhat different system can be used. An array of transmitters at frequencies f_1, f_2, and f_3, together with their respective receivers, would permit intensity-modulated oscilloscope traces to be generated, the appearance of which would be similar to the curves of Fig. 8d. Such an arrangement would be quite elaborate, but a simplified arrangement could be made. A transmitter at frequency F_o could be modulated at frequency ΔF_o, $2\Delta F_o$, $3\Delta F_o$, ... simultaneously. A single transmitter with such modulation, therefore, would correspond to a transmitter operating with the sidebands f_1, f_2, f_3, ... (see Fig. 9). The receiver would apply the received signal after detection and amplification to the separate tuned amplifiers ΔF_o, $2\Delta F_o$, and $3\Delta F_o$ (see Fig. 9) and then the display will consist of an intensity-modulated trace from each sideband-tone amplifier.

In another arrangement, the transmitter would sweep through the requisite frequency range (f_1 to f_3 for the example shown in Fig. 8) corresponding to the biological-voltage range. A receiver slaved to the transmitter would deliver a constant output when no implanted device resonance was present in the frequency range f_1 to f_3. When there was a resonance in this frequency range, the different magnitudes of RF received and amplified at the receiver would cause the generation of a pulse, the duration of which would be adjusted to equal the transmitter/receiver sweep period. The pulse height that would be the output of an FM discriminator would be proportional to the frequency. Special circuits could smooth the output so that the recovered waveform (see Fig. 8e) was presented. This method, by utilizing a simultaneously swept receiver and transmitter, would be used for passive telemetry of variables varying slowly with time such as temperature, pressure, or pH. In the case of pressure and temperature telemetry, the implanted device would consist of an L and C, one of which would be pressure or temperature sensitive. In the case of pH, a set of pH electrodes with a voltage range of 60-600 mv could be used to indicate a pH voltage of 1-10 mv.

Another arrangement, shown in Fig. 10, would continuously track the frequency of the implanted device resonant circuit by means of a scanning





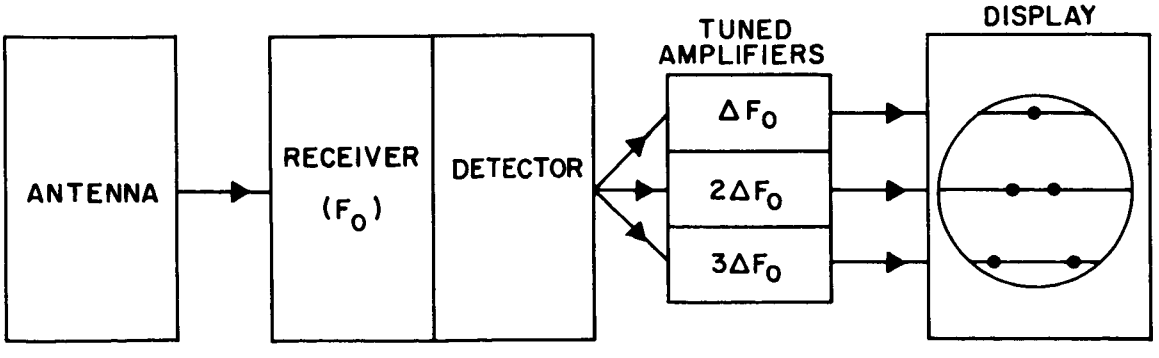


Figure 9. Sideband Telemetry System

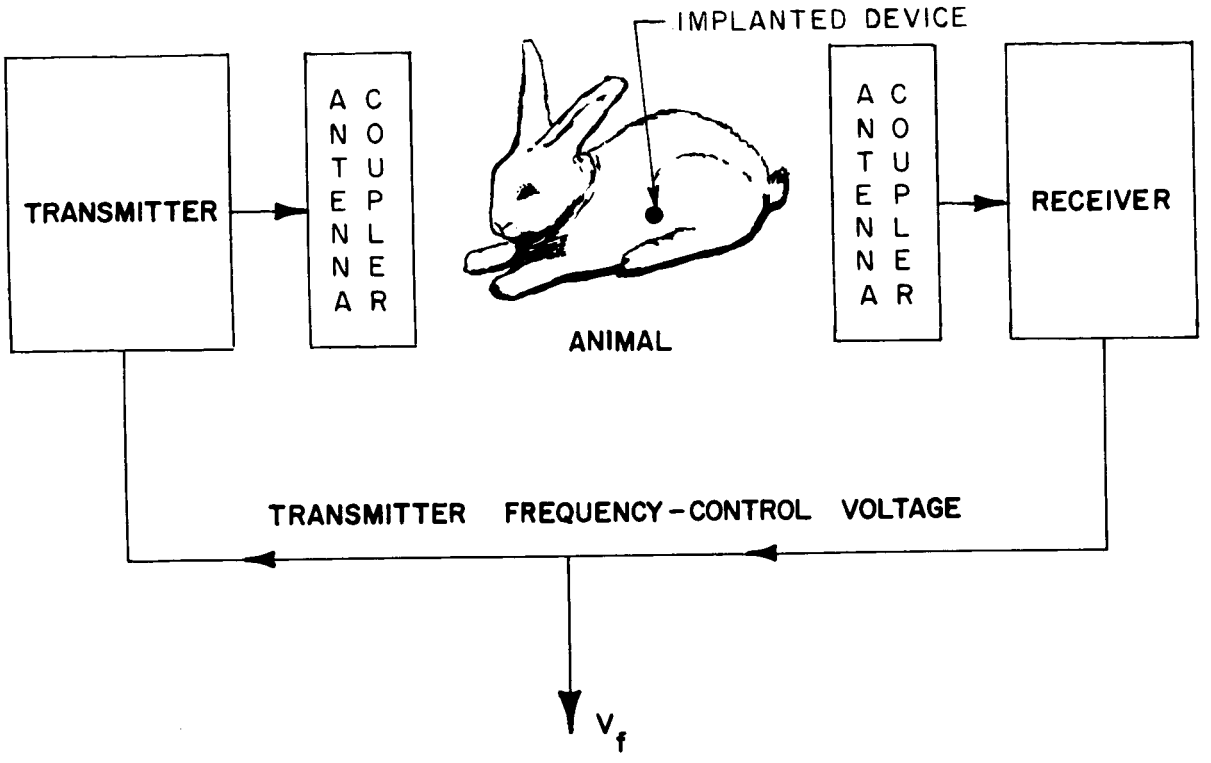


Figure 10. Frequency-Tracking Telemetry System

transmitter/receiver arrangement capable of lock-on and reduced scan. The correction voltage V_f would control the transmitter's center frequency and would be proportional to the resonant frequency of the implanted device.

The above methods for voltage telemetry will generate a continuous record of a voltage that is a function of the original biological voltage. An accurate reproduction of the original waveform V_b, shown in Fig. 8a, can be derived from an inverse transformation of V_R in Fig. 8d using Eqs. 24 and 25 so that

$$V_R = s\,f \tag{26}$$

$$V_b = \frac{1}{100}\,(V_1) \tag{27}$$

where s is a proportionality constant.

The result of such a transformation is

$$V_b = u\,(V_R)^4 + w \tag{28}$$

where u and w are proportionality constants. This transformation can be accomplished with relative ease by using two squaring circuits for which u may be established by Eqs. 24-27.

A means of transformation that might be simpler and more accurate takes into account departures from Eqs. 24 and 25, nonlinearities in transformer T_1, and the L of the implanted device (see Fig. 11). An additional device is used in the external equipment. Nonloading leads are attached to the RF-tuned circuit,

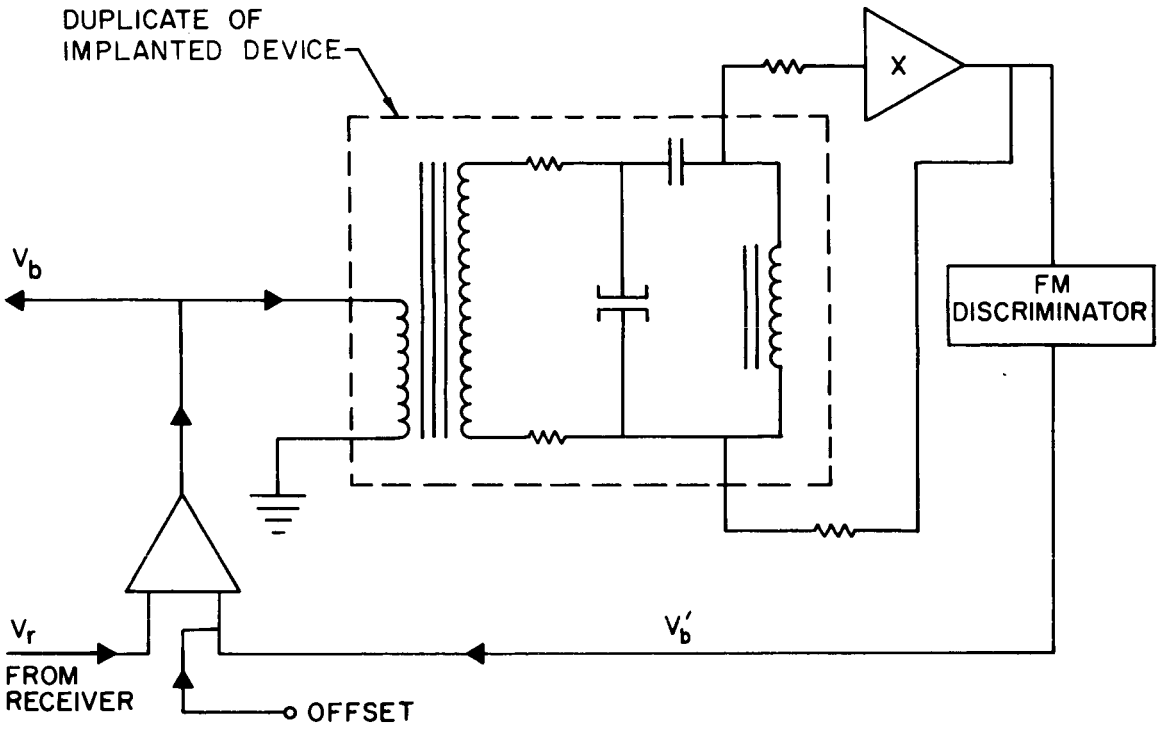


Figure 11. Biological Restoration System

which forms the frequency determining part of a feedback oscillator using amplifier X. The output of this amplifier/oscillator goes to an FM discriminator and generates a synthesized voltage V_b'. This voltage is fed to a difference amplifier together with the V_r available from the receiver. The voltage difference V_b drives the external device and determines the frequency of the RF circuit. If the correct polarities and dc offsets are effected, the system should drive V_b so that it is the restored biological voltage.

EXPERIMENTAL RESULTS

On the basis of published work [29-31], assuming that the RF employed is less than 100 Mc, the dielectric constant of tissue K is approximately equal to 80, which corresponds to the dielectric constant of water. Further, assuming that the conductivity of tissue is similar to that of sea water, which has a resistivity of 50 ohm-cm [23], initial tests of the passive telemetry system were carried out using sea water as a substitute for tissue. Simulated biological-voltage sources with biological impedances equal to the values given earlier for tissue (1,000-2,000 ohms) were used.

The telemetry device and biological simulator were tested for RF range and propagation and to determine the maximum operating depth in tissue. The choice of operating frequency was not critical, but, from a consideration of ferrite-loaded loops [29], electrically small antennas and inductors [24, 26], and operation in tissue [30, 31], the 5-10 Mc region was selected.

It has been shown that the field of a dipole-radiator enclosed in an insulating sphere immersed in a conducting medium provides dipole radiation fields [30]. The design of the antenna in the device was governed by omnidirectional considerations. The previous work on this subject [11, 25], together with the special considerations necessary for this method of passive telemetry, dictated the use of two mutually perpendicular coils.

A simulated tissue depth of 6 in. was achieved, but, because operation at greater depths was not attempted, 6 in. might not represent the tissue-depth limit. Transmitter power was approximately 0 dbm. Figure 12 is a photograph of the assembly and parts of the implanted device, the circuit diagram of which is shown in Fig. 7.

MICROMINIATURE DEVICES

The passive telemetry device previously described requires about 10,000 cells and voltages of 1-10 mv for operation. It would be valuable to have a device capable of operating with 10-100 cells and with electrode voltages as low as 50-100 μv. It may be, however, that better electrode material or design would make the full-cell transient voltage of 100 mv available from such a small number of cells. The internal impedance of the cell tissue would lie in the range of 10^6 -10^5 ohm

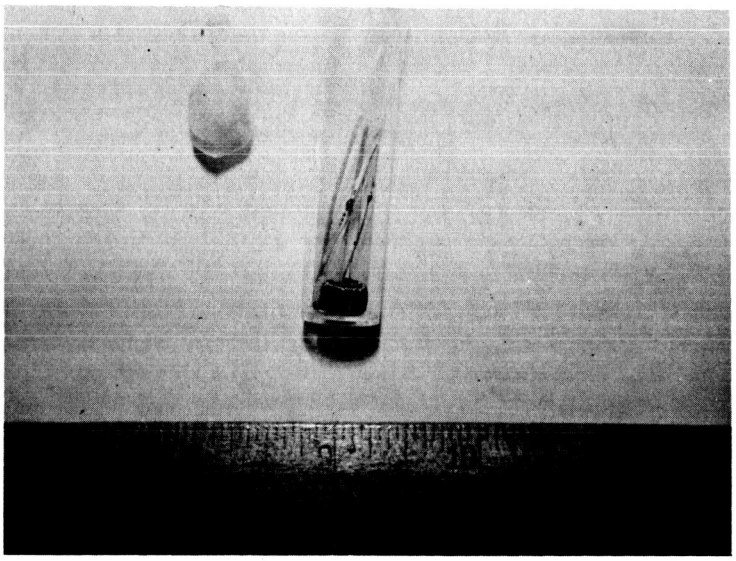


Figure 12. Assembled Implanted Device and Parts

for 10-100 cells touching the electrode. Passive telemetry from such a small number of cells might be called "microscopic" passive telemetry to distinguish it from the previous work, which might be called "macroscopic" passive telemetry.

A "microscopic" passive telemetry device operating with a small number of cells should have the following advantages:

1) Small size: a cylinder 0.030 in. in diameter and 0.030 in. high
2) Capability of bipolar-voltage indication from tissue in the 50-100 μv range from tissue impedances of 10^6-10^5 ohms
3) Capability, because of its small size, of being injected into place without greatly disturbing the animal

Possible methods by which such a telemetry system might operate are as follows (all of these methods are inherently bipolar because of the linearity of the frequency change with video voltage):

1) Electrostrictive control of an acoustic resonator frequency
2) Magnetostrictive control of an acoustic resonator frequency
3) Modification of the capacitor described by the curve of Fig. 6 and Eq. 24 for back-biased junction diodes so that the largest percent change in capacitance occurs for extremely small voltage changes about zero volts

Item 3 above will permit the 100:1 step-up transformer to be eliminated. It is needed in the present passive "macroscopic" system because an appreciable capacitance change does not occur in the voltage range from 0 to 0.01 v of back bias [32]. Another reason why it is needed in the present system is that the level of RF energy should cause lower RF voltages to exist at the voltage-variable capacitor than video voltages so that parametric operation with respect to the RF does not occur. In other words, the RF should not vary the capacitance of C_v.

Items 1 and 2 above represent the most efficient solutions, however, because telemetry systems using such methods would be able to have implanted devices much simpler and smaller than the present "macroscopic" device. No separate R, L, or C components would be required. The implanted devices would consist simply of a cylinder vibrating in the longitudinal mode at RF frequencies in the 5-18 Mc range with a means for having the biological voltage vary this frequency.

On the basis of published work [33-36], operation of a cylindrical acoustic resonator of either type with dimensions of 0.030 in. in diameter and 0.030 in. high might be feasible. External RF at the resonant frequency of the cylinder would induce acoustic resonance, which would occur directly in the magnetostrictive resonators but would require provisions for RF coupling in the case of the electrostrictive resonator. The tissue voltages would vary the electric field at a video rate in a portion of the volume of the electrostrictive resonator (or vary the magnetic field, in the case of the magnetostrictive resonator). Because the Q of the resonant cylinder might be made quite high (beyond 50,000 and 20,000 for the electrostrictive and magnetostrictive resonators, respectively), large changes in the externally radiated and reflected power will occur for extremely small frequency deviations of the cylinders.

Figure 13 shows a comparison of the dimensions for the "macroscopic" device and the "microscopic" device. The "microscopic" device will have two leads that can be used to generate an electric field directly, for the electrostrictive case, or can be run to a coil enclosing the cylindrical resonator, for the magnetostrictive case. These two leads are the electrode leads.

As shown in Fig. 13, the "microscopic" device has the cylindrical resonator mounted inside a rigid RF-transparent capsule that serves as a base for one end of the cylinder and a cover that will maintain the other end free. This arrange-

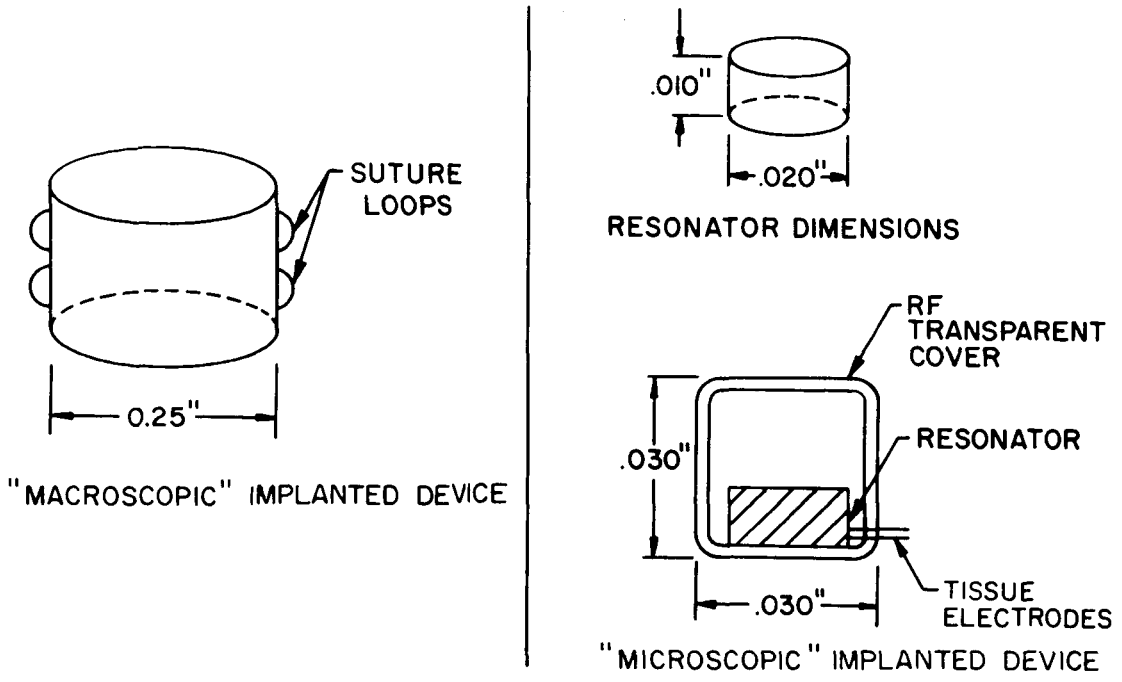


Figure 13. Dimensions for Implanted Devices

ment is a method for using a fundamental acoustic mode; other acoustic modes are possible, each providing operation that can be adapted for different conditions.

Because the Q is quite high and the resonators can be fabricated in a similar manner to that of present-day piezoelectric quartz crystals, the center frequency can be specified, and it can be "built in" quite accurately. This fact might permit many "microscopic" devices to be injected into a given animal. External transmitter/receiver arrangements of relative sophistication should be capable of presenting simultaneous multipoint information, and the center frequency will identify the site of each of the "microscopic" devices.

In the case of the "microscopic" devices particularly, but also in the case of the "macroscopic" devices, some additional consideration must be given to electrode material and design for long life and minimal polarization. Quite a bit of laboratory and clinical work that is oriented exclusively to the problem of electrode designs that will preserve the integrity of the electrode-tissue connection for the lifetime of the animal must be performed. Electrode depolarization and tissue reaction at the electrode-tissue connection must be minimized for the reliable operation of the passive telemetry device.

APPENDIX

DESIGN OF STEP-UP TRANSFORMER

The design of the step-up transformer using the equivalent circuit shown at the top of Fig. 14 can be accomplished with the following approximations:

$a = 1/100$

$R_g \simeq 1000$ ohm

$R_L > 10^8$ ohm

$C_S < 10\ \mu\mu f$

L_S, R_c, L_P', R_p are negligible

$R_S << R_L$

The above approximations result in the equivalent circuit at the bottom of Fig. 14.

To minimize the over-all size of the transformer and to permit a relatively high turns ratio, the transformer can be constructed with the following features:

1) The toroidal core with a tape-wound coil on a stainless-steel bobbin can have a maximum diameter of 0.280 in. and height of 0.160 in. (e.g., Magnetics, Inc. Bobbin No. 80535 and 80523).

2) A one-mil tape of superalloy with 20-40 wraps permits initial permeability in the range of 50,000 to 150,000 at 100 cps.

3) The secondary winding can consist of 4,500 to 7,000 turns of No. 46 to No. 52 single Formvar wire. (This is a difficult feat requiring much care in fabrication, but the finished winding is relatively rugged after potting, although care must be exercized so that poisoning of the superalloy does not occur. Superalloy is strain sensitive, but the stainless-steel bobbin provides strain protection).

4) The primary winding with 1/100 the number of turns of the secondary specifies L_p in Fig. 14. For toroidal cores

$$L_p = \frac{1.25\ N_i^2\ A_c\ M_o \times 10^{-8}}{\ell_g} \qquad (29)$$

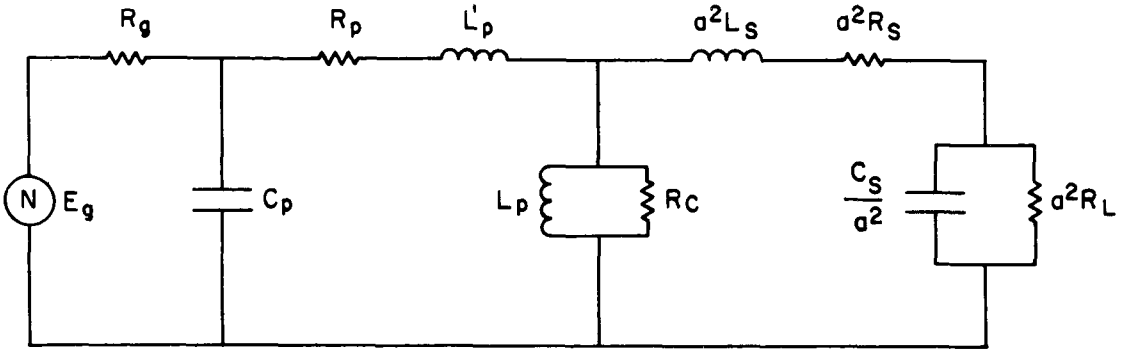
where:

N_i = number of turns of coil

A_c = cross-sectional area of core

ℓ_g = length of magnetic path (mean circumference)

M_o = initial permeability of toroid



a = TURNS RATIO = N_p/N_S

C_p = PRIMARY EQUIVALENT SHUNT CAPACITANCE

C_S = SECONDARY EQUIVALENT SHUNT CAPACITANCE

E_g = GENERATOR VOLTAGE

R_p = PRIMARY WINDING RESISTANCE

R_S = SECONDARY WINDING RESISTANCE

L_p = PRIMARY INDUCTANCE = MAGNETIZING INDUCTANCE

L'_p = PRIMARY LEAKAGE INDUCTANCE

L_S = SECONDARY LEAKAGE INDUCTANCE

R_C = CORE LESS EQUIVALENT SHUNT RESISTANCE

R_g = GENERATOR IMPEDANCE

R_L = LOAD IMPEDANCE

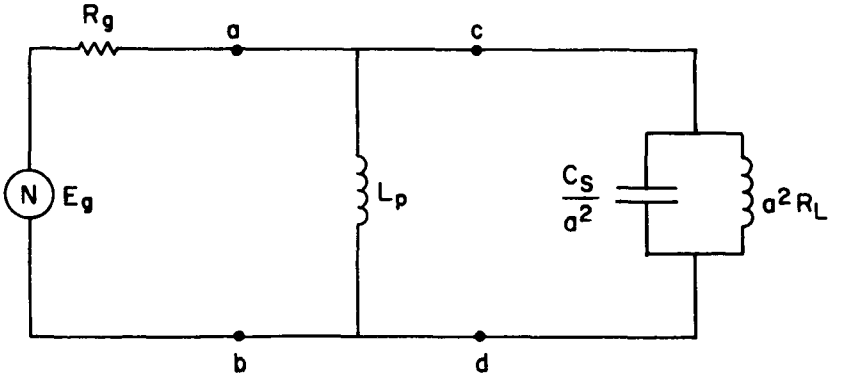


Figure 14. Schematic of Video Step-Up Transformer

If the secondary consists of 4,500-7,000 turns, then a 1/100 turns ratio requires 40-70 turns for the primary. The value of L_p for miniature tape-wound cores is in the range of 15-20 mh. The value of X_{L_p} is approximately 120 ohm at 1,000 cps. The value of the voltage across the primary of the transformer is about one tenth that of the voltage E_g. Because the reactance of L_p will vary linearly with frequency, loading of the tissue by the primary magnetizing inductance will increase at frequencies lower than 1,000 cps. Indications are that the voltage E_g available from heart tissue might be as large as 10-15 mv. To bring this value up to more than 500 mv, the turns ratio can be changed to less than 50:1 so that the primary magnetizing inductance is at least four times its previous value (500 ohm).

Because the voltages and currents associated with the action potential spikes in single cells are about 100 mv and 10^{-8} amp and the impedance is 10^{7} ohm, the sum of the currents of 10,000 such cells is $10^{4} \times 10^{-8} = 10^{-4}$ amp = 100 μamp.

The magnetizing force caused by this current in the primary of the transformer is:

$$H = 1/2\left[\frac{(10^{-4})\ N_i}{\ell_g}\right] \qquad (30)$$

where ℓ_g, the length of the magnetic path in inches, is 0.23 in. for the cores under consideration. Letting $n_i = 1$

$$H = 1/2\left[\frac{10^{-4}}{0.23}\right] = \frac{10^{-4}}{0.46} \cong 2 \times 10^{-4} \text{ oersted per turn}$$

$$H = 10^{-2} \text{ oersted per 50-turn primary}$$

The coersive force for superalloy at the following frequencies is:

dc		0.003-0.009
60	~	0.02-0.06
400	~	0.03-0.10

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